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Coupled motions in human and porcine thoracic and lumbar spines

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ABSTRACT

Coupled motions, i.e., motions along axes other than the loaded axis, have been reported to occur in the human spine, and are likely to be influenced by inclined local axes due to the sagittal plane spine curvature. Furthermore, the role of facet joints in such motions is as yet unclear. Therefore, this study aimed at assessing coupled motions in multiple spine sections *in vitro*, before and after removal of posterior elements. Six elderly human and 6 young porcine spines were sectioned in four segments (high thoracic, mid thoracic, low thoracic and lumbar), each consisting of four vertebrae and three intervertebral discs. Segments were loaded along each of the three axes, and three-dimensional rotations of the middle segment were quantified. Subsequently, posterior elements were removed and the protocol was repeated. To avoid mixed loading between Axial Rotation (AR) and Lateral Bending (LB), in contrast to other studies, local axes at the vertebrae were defined as aligned with the loading device prior to each load application.

Expressed as a percentage of motion in the loaded direction, coupled motions were on average larger in human (22.7%, SD = 2.2%) than in porcine (11.9%, SD = 1.2%) spines ($p < .001$). Largest coupled motions were obtained in AR loading of the lumbar spine segments, with mean magnitudes averaged over coupling axes for human L2-L3 joints of 48.9% (SD = 13.2%), including somewhat more LB (56.4%, SD = 18.6) than FE (41.4%, SD = 14.1%) coupling. For porcine L3-L4 joints average coupling in AR loading was 29.3% (SD = 8.2%). In human segments removal of posterior elements only had substantial effects in the lumbar spine segments, where posterior element removal decreased coupled motion during AR loading, averaged over LB and FE coupling, from 48.9% (SD = 13.2%) to 27.7% (SD = 6.1%), mainly through increased motion in the loaded direction.

The present results indicate that coupled motions were largest in the lumbar spine. In human spines, posterior elements only contributed to coupled motions in lumbar axial rotation loading.

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1. Introduction

During intended uni-axial *in vivo* motions, the human spine has been shown to generate substantial so-called coupled motions in the thoracic (Fujimori et al., 2012; Fujimori et al., 2014) as well as the lumbar (Fujii et al., 2007; Gercek et al., 2008; Li et al., 2009; Ochia et al., 2006; Shin et al., 2013) spine. These coupled motions suggest that loading about a single axis causes rotations about more than one axis. Moreover, enhanced *in vivo* lumbar coupled motions have been associated with low back pain (Cheng et al., 2013; Lund et al., 2002) and degenerative disc disease (Li

et al., 2011), and could thus be indicative of spine problems. However, a difficulty in interpreting *in vivo* data in terms of kinematic coupling is that, when asked to perform a uni-axial motion, subjects, and maybe especially patients, may not apply pure uni-axial loads to their spine (Grabiner et al., 1992; Shirazi-Adl, 1994b).

In vitro as well as in modelling, pure uni-axial moments can be applied, which allows for a more direct interpretation of anatomically-induced kinematic coupling. *In vitro* experiments have also shown coupled motions in the thoracic (Mannen et al., 2015; Panjabi et al., 1976) and lumbar (Oxland et al., 1992a; Panjabi et al., 1989, 1994) spines and the thoracolumbar junction (Oxland et al., 1992b). Facet joints have been suggested to play an important role in this coupling (Oxland et al., 1992a; Panjabi et al., 1989).

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However, even *in vitro*, where the loading direction can be controlled, interpretation of data requires consideration. In most studies referenced above (Oxland et al., 1992a; Panjabi et al., 1989, 1994), sections of multiple spine segments were loaded. Subsequently, unlike modelling work where vertebral motions were expressed in global axes (Shirazi-Adl, 1994a,b), local axes systems were constructed on vertebrae and motions of vertebrae were expressed relative to their neighbour. While the latter seems appropriate from an anatomical perspective, a problematic aspect is that the local axes are not necessarily aligned with the loading direction. This alignment problem has been acknowledged as well in previous *in vitro* work (Oxland et al., 1992a; Panjabi et al., 1989), and, through modelling work, has been shown to partly explain *in vitro* findings (Cholewicki et al., 1996). To clarify the problem, Fig. 1 shows axes systems in a lordotic spine alignment, which is normal for the lumbar spine. This figure shows that with inclined local axes, moments along the lateral bending or axial rotation axis of a loading device can project on local axial rotation or lateral bending axes, respectively. This projected moment differs in sign between the upper and lower vertebra if they are inclined in opposite directions. Indeed, coupled motions reported from tests on multi-segmented lumbar spines (Oxland et al., 1992a; Panjabi et al., 1989, 1994) are generally consistent with such projections due to tilting of vertebrae resulting from a normal lumbar lordosis. In this respect, Oxland et al. (1992a) and Cholewicki et al. (1996) distinguished postural and structural/mechanical coupling. The alignment problem might lead to interpretation problems if elderly subjects, or subjects with low back pain have a more pronounced spinal curvature, because in that case coupled motions expressed in local axes are expected to be larger. To avoid moment projections being interpreted as (mechanical/structural) kinematic coupling, one could argue that local axes should, prior to loading, be aligned with the loading device.

Following this line of reasoning, the current study set out to assess coupled motions in human thoracic and lumbar spine sections *in vitro*, with local axes at onset of loading aligned with the

loading device. We hypothesized that coupled motions would be smaller after posterior element removal, and differ between lumbar and thoracic spine sections. In addition, animal models such as porcine spines are commonly used as a proxy for human spines. Similarities as well as substantial differences with human spines have been noted for both anatomy (Busscher et al., 2010a) and biomechanical behaviour in the loaded direction (Wilke et al., 2011). As coupled motions have not yet been compared between human and porcine spines, we additionally assessed to what extent coupled motions in porcine spines are roughly similar to elderly human spines.

2. Materials and methods

2.1. Specimen and specimen preparation

Fresh frozen spines of six human and six porcine cadavers were used in this study. The human spines were harvested from six cadavers obtained from the Department of Anatomy of the University Medical Center Groningen, The Netherlands. Mean age of the subjects at the time of death was 72 years (range 55–84 years). The porcine spines (from about 40 kg domestic pigs) were obtained from a local abattoir. All spines were dissected from T1-S1 and all musculature was carefully removed, while leaving the ligaments, facet joints, and joint capsules intact. At both sides of the spine, approximately 3 cm of the ribs was preserved, including the costovertebral and costovertebral articulations. CT scans of the spines showed normal porcine spines and mildly degenerated but otherwise normal human spines (Busscher et al., 2009).

The spines were divided into high-, mid-, low-thoracic, and lumbar segments, each containing 4 vertebrae and 3 intervertebral discs. The porcine spines had 15 or 16 thoracic and 6 lumbar vertebrae, compared to 12 and 5, respectively, in the human spines. The porcine spines were dissected in T2-T5, T7-T10, T12-T15, and L2-L5. The human spines were dissected in T1-T4, T5-T8, T9-T12, and L1-L4.

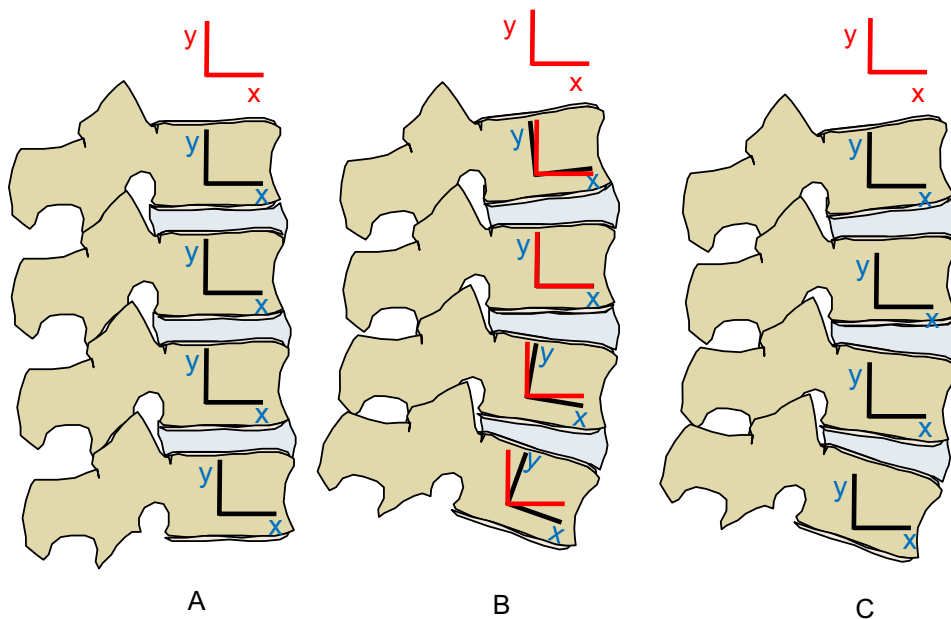


Fig. 1. Sagittal plane illustration alignment effects. With all vertebrae initially aligned with the loading device (A), local axes (black) coincide with those of the loading axes (red). In a lordotic spine section, local anatomy-based axes do not coincide with the loading device (B). Consequently, moments along the red x or y axis will partly project on the black y or x axis, respectively. For example, a positive (right Lateral Bending) moment along the red x would cause a negative (left Axial Rotation) moment along black y of the top vertebra, and a positive (right Axial Rotation) moment along the bottom vertebra black y. Note that in (B), dependent on Euler decomposition order, the problem is further exacerbated because the black axes of adjacent segments are also not initially aligned. In the present study, these problems are avoided by aligning local axes with the loading device prior to each load application (C). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

The specimens were stored at -40°C . After thawing and removal of musculature, the upper and lower vertebrae of the segments were embedded in metal cups fitting the test set-up using a low melting temperature alloy (Cerrolow-147; 48% bismuth, 25.6% lead, 12% tin, 9.6% cadmium, and 4% indium). Three screws were driven into the upper and lower endplate to optimize the connection between the vertebrae and the alloy. Embedding and screws did not restrict facet joint or intervertebral disc motion. The top vertebra was embedded first, and the lower vertebra was then embedded, while the spine segments were hanging freely in order to obtain natural alignment. A pin containing three LED's was rigidly fixed to the anterior surface of each vertebral body. To minimize dehydration, a saline-soaked gauze was wrapped around the spine sections and they were sprayed with saline solution during preparation and testing. All tests were performed at room temperature.

2.2. Instrumentation

Before testing, an axial compressive preload of 250 N was applied to both the porcine and human specimen for 1 h to simulate physiological conditions and to reduce effects of superhydrated intervertebral discs (Emanuel et al., 2015). During subsequent application of moments, no compression was applied to avoid buckling. Because testing of porcine and human spines was performed at different locations, the driving machine differed between species. The testing device (see below) was driven by a Zwick material testing system (model TC-FR2.5TN, Zwick Roell, Ulm, Germany) for the tests with human segments and an Instron material testing system (model 8872, Instron&IST) for the tests with porcine segments. The displacement rate was set at $0.5^{\circ}/\text{s}$.

Motions of the LED's were recorded by an optoelectronic 3D movement registration system with one array of 3 cameras (Optotrak 3020, Northern Digital Inc, Waterloo ON), and synchronized with the loading system. The accuracy of this system is better than 0.05 mm. The sample rate was 50 samples/s. The global axes of the Optotrak system were aligned with the loading device by placing the system calibration block along the edges of the base plate of the loading device.

A custom-made loading device was used in this study (Busscher et al., 2010b). In this device, flexion-extension and left-right lateral bending are applied using four-point bending. It should be noted that our test setup always included four vertebrae. Consequently, whereas the middle two vertebrae, which were used to quantify local coupled rotations, could move in any direction, the embedded top and bottom vertebrae were not free to move in all directions. For FE loading, LB was blocked at both ends but the top embedding cup could axially rotate around its centre. For LB loading, the specimen was axially rotated 90° . Therefore, FE was blocked at both ends and AR was again free. For AR loading, FE and LB were blocked at both ends and the AR was imposed using a steel cable rotating the top cup. Furthermore, in all cases, the bottom vertebra could translate in the axial direction using a low-friction sliding rail.

2.3. Loading protocol

For the human spines, three cycles from -4 N m to $+4\text{ N m}$ were applied in each of the three planes of motion, flexion-extension (FE), lateral bending (LB) and axial rotation (AR). Because of the smaller size and larger flexibility, similar loading but with cycles from -2 N m to $+2\text{ N m}$ was applied to the porcine spines. Subsequently, posterior elements were removed and loading was repeated. Furthermore, unrelated to the present analysis, creep testing of 30 min at 50% of the cyclic load levels, followed by 30 min of recovery and removal of interspinous and intertransverse ligaments, preceded and followed by sets of 3 load cycles as

described above, had been applied prior to the test analyzed here (Busscher et al., 2010a). Creep was shown to have minor effects ($<10\%$) on loaded direction Range of Motion (ROM), with full recovery after the recovery period in all directions except lateral bending (Busscher et al., 2011); ligament removal had hardly any effects on ROM (Busscher et al., 2011). The present study used data just prior to and immediately following posterior element removal, so that preceding tests are to be considered as preconditioning.

2.4. Data analysis

A custom Matlab (Mathworks, Natick MA, USA) program was used to analyse the data. At each instant of time (t) during measurements, local axes ($R_{i,t}$) were constructed on each vertebra (i) using the position of the three non-collinear LED's. The first 10 samples of each measurement during which no load was applied, were considered as a reference ($R_{i,\text{ref}}$). The transpose of this matrix ($R_{i,\text{ref}}^T$) was used to obtain, for each vertebra at each instant of time, its orientation relative to the unloaded condition, according to:

$$R_{i,t,\text{aligned}} = R_{i,t} \cdot R_{i,\text{ref}}^T$$

in other words, local axes were aligned with the loading device prior to load application, as in Fig. 1C. Subsequently, the orientation of each vertebra was expressed relative to its caudal ($i-1$) neighbour:

$$R_{i,i-1,t} = R_{i-1,t,\text{aligned}}^T \cdot R_{i,t,\text{aligned}}$$

Finally, the angular motion of each of the three intervertebral joints was obtained by Euler decomposition. The decomposition always started with the loaded axis (Mannen et al., 2015), using the order: FE-LB-AR for FE loading, LB-FE-AR for LB loading, and AR-FE-LB for AR loading. In this way, for each of the loading directions, angular motions about all three axes of motion were calculated for each of the three intervertebral joints. Further analyses were applied to the unconstrained middle intervertebral joint only. Angles at positive and negative peak loads were calculated, both for the loaded direction and for the two secondary (coupled) axes. Furthermore, absolute ROM values were calculated for each axis, and the ranges along the secondary axes were expressed as percentage of the average ranges in the loaded planes (coupling percentage).

2.5. Statistics

To allow for inclusion of all loading directions within a single ANOVA per species, coupling percentages were averaged across coupling axes. In the absence of major deviations from normal distribution, a three-way ANOVA was applied (IBM SPSS version 23), separately for the human and porcine spines, with spine as between subject factor, and with posterior element removal (pre and post), loading direction (FE, LB, AR), and spine level (High Thoracic, Mid Thoracic, Low Thoracic and Lumbar) as within subject factors. As load levels and spine sections were not equal across species, the comparison between human and porcine spines was only made for coupling percentages after averaging over all spine levels and loading directions, using a non-paired t -test. Effects were considered significant when $p < .05$.

3. Results

In both human and porcine spine sections, 6 out of 72 (24 sections \times 3 loading directions) pre/post comparisons (concerning 2 human upper thoracic and 4 porcine thoracic sections from different levels) were discarded, mainly because fracture occurred at some point during loading after posterior element removal. Abso-

Table 1

P-values for effects of posterior element removal, spine level and loading direction on coupling percentages, based on separate ANOVA's for human and porcine spine sections.

	Human	Porcine
Prepost	.193	.084
Level	.001	.001
Direction	.001	<.001
Level * direction	<.001	<.001
Prepost * direction	.077	<.001
Prepost * level	.013	.292
Prepost * level * direction	.033	.026

lute coupling percentages, averaged over all spine sections, loading directions, and pre- and post-removal of posterior elements, were larger in human (22.7%, SD = 2.2%) than in porcine (11.9%, SD = 1.2%) spines ($p < .001$).

Absolute coupling percentages (averaged over the two coupling axes), varied with loading direction and spine level, for both human and porcine spines (Table 1, Figs. 2 and 3). In human spines, those percentages were highest for AR loading of the lumbar section (average [48.9%, SD = 13.2%], which included somewhat more LB [56.4, SD = 18.6] than FE [41.4%, SD = 14.1%]). In the other 11 spine section/loading direction combinations, coupling percentages ranged from 14.0% (SD = 2.5%) for LB loading of the lumbar section to 29.3% (SD = 8.2%) for FE loading of the mid thoracic section, prior to posterior element removal. In porcine spines, percentages coupling ranged from 7.6% (SD = 1.1%) for FE loading of the low thoracic section to 25.9% (SD = 5.8%) for AR loading of the lumbar section.

The overall averaged percentage coupling was not significantly different after (22.3%, SD = 2.5%) compared to before (23.2%, SD = 2.2%) posterior element removal (Table 1), and the same was true for porcine spines (12.0%, SD = 1.2% vs. 11.9%, SD = 1.3%). However,

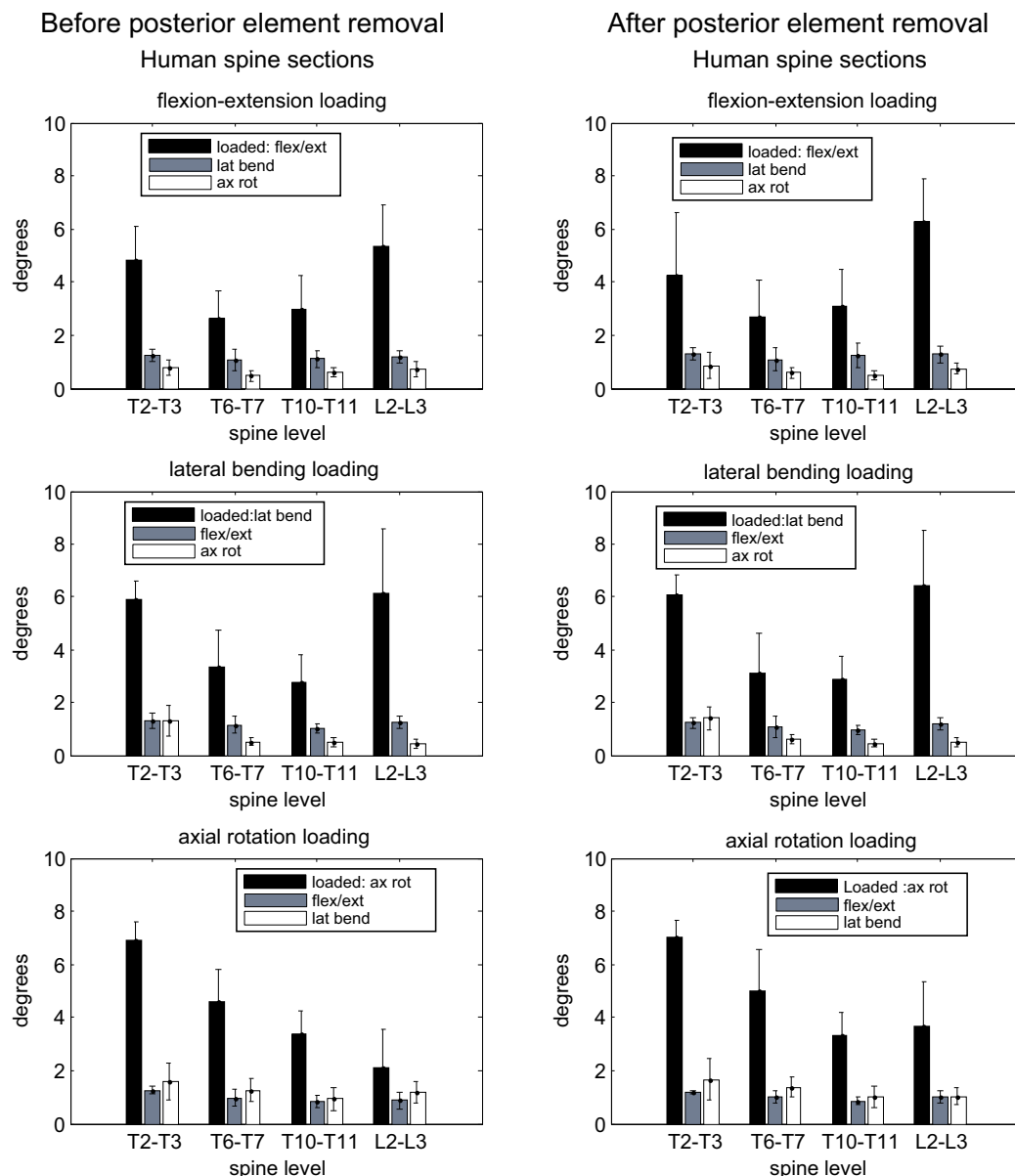


Fig. 2. Three-dimensional range of motion of human spine sections during 4 N m flexion-extension (top row), lateral bending (middle row) and axial rotation loading (bottom row), before (left column) and after (right column) removal of posterior elements. All spine sections consisted of 4 vertebrae, and the analyzed data concern the middle intervertebral joint. The segment level of this joint is indicated on the x-axis. Error bars indicate one standard deviation.

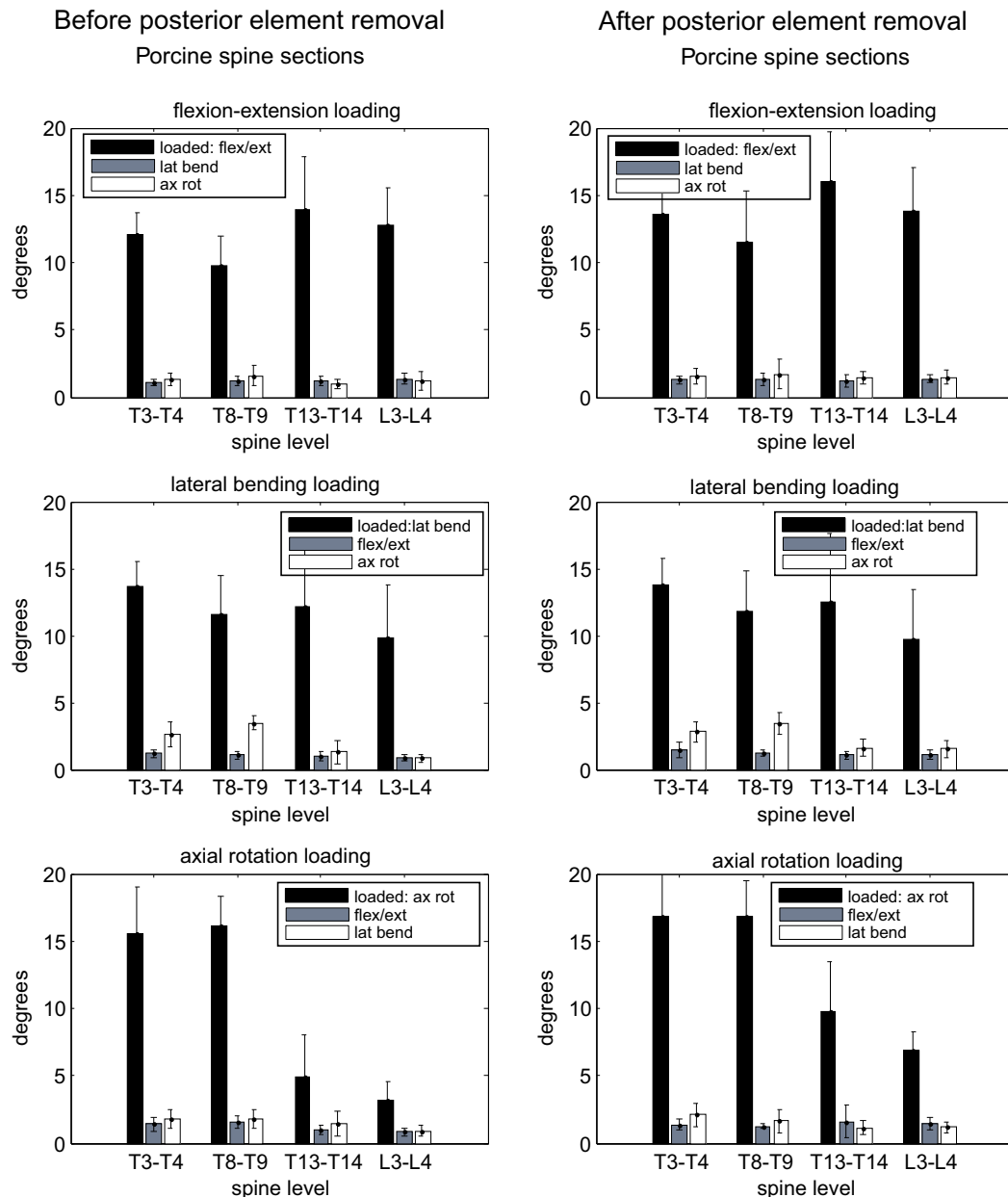


Fig. 3. Three-dimensional range of motion of porcine spine sections during 2 N m flexion-extension (top row), lateral bending (middle row) and axial rotation loading (bottom row), before (left column) and after (right column) removal of posterior elements. All spine sections consisted of 4 vertebrae, and the analyzed data concern the middle intervertebral joint. The segment level of this joint is indicated on the x-axis. Error bars indicate one standard deviation.

a significant two-way interaction with spine level (human), loading direction (porcine) and a three-way interaction with both spine level and loading direction (human and porcine) was found (Table 1; Fig. 4). In human spines, this was mainly caused by a significant ($p < .001$) reduction of coupling, averaged over coupling axes, from 48.9% (SD = 13.2%) before, to 27.7% (SD = 6.1%) after posterior element removal for AR loading of lumbar sections. A closer inspection showed that both left and right AR loading resulted in a change towards more extension coupling with 0.55° (SD = 0.48° ; paired t -test $p = .04$) and 0.8° (SD = 0.44° ; paired t -test $p = .01$), respectively, after posterior element removal. In lumbar sections of porcine spines, coupling after posterior element removal was reduced for AR loading (from 27.7%, SD = 5.8% to 19.1%, SD = 1.9%; $p = .009$) but increased for LB loading (from 9.0%, SD = 2.2% to 13.7%, SD = 3.9%; $p = .023$).

While the mean and standard deviation of percentages coupling range of motion showed a quite consistent pattern, it should be noted that individual curves for coupled motions, while being highly reproducible in repeated load cycles, were mostly neither smooth, nor monotonous.

4. Discussion

We found substantial coupled motions, which, in line with our hypotheses, depended on spine level and loading direction. Consistent with previous reports (Panjabi et al., 1989, 1994), coupled motions were largest for axial rotation loading of the lumbar spine.

By aligning local axes with the global axes prior to each load test, we avoided axes systems that are not aligned with the loading

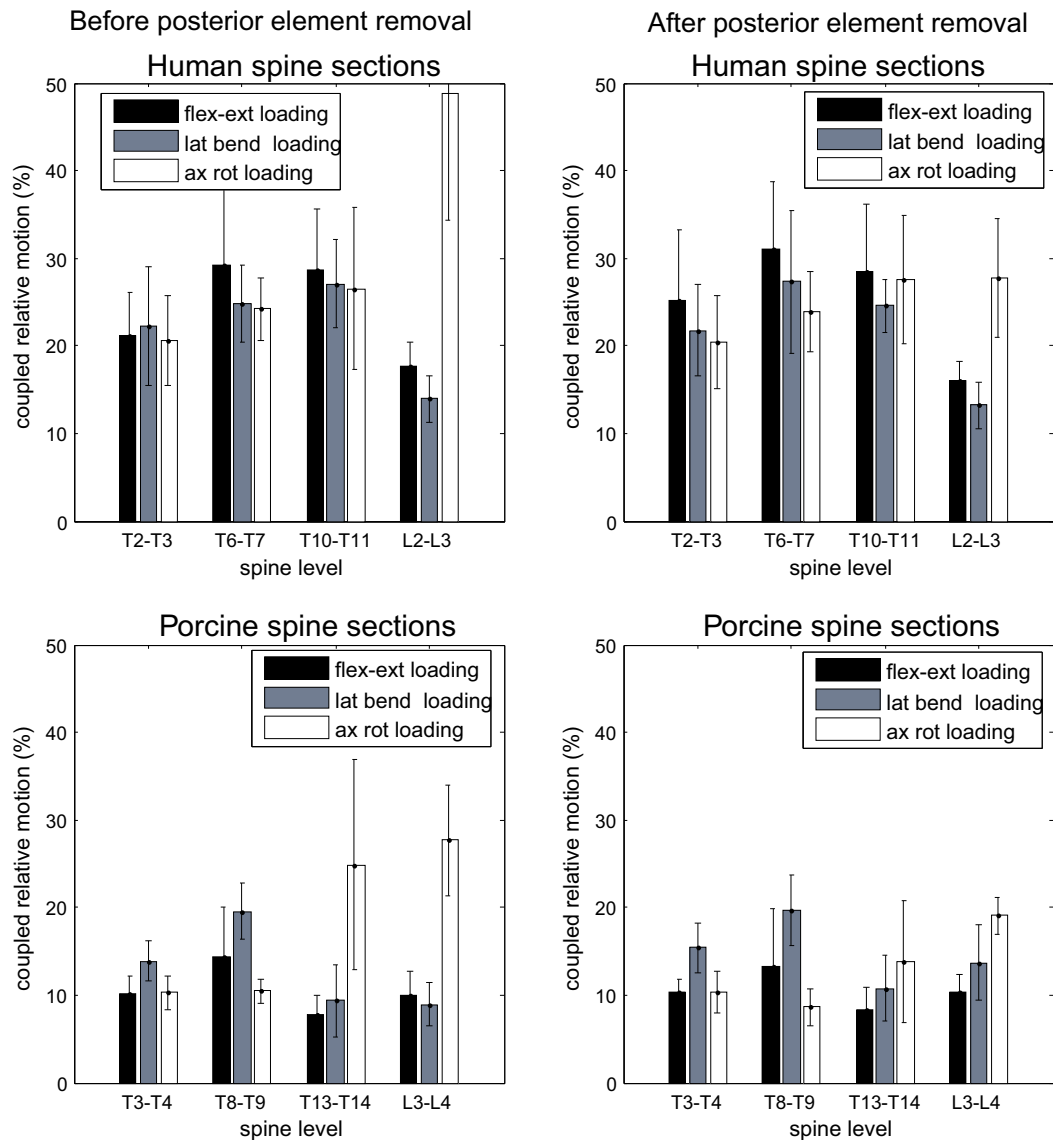


Fig. 4. Coupling percentages (averaged over coupling axes, and expressed as percentage of average motion in the loaded direction) for human (top row) and porcine (bottom row) spines before (left column) and after (right column) removal of posterior elements.

axis and not aligned between subsequent vertebrae (Hof et al., 2001). Consequently, our results should represent effects of mechanical/structural coupling only, and not combined postural and structural/mechanical coupling (Cholewicki et al., 1996; Oxland et al., 1992a), which may have affected previous *in vitro* and *in vivo* studies.

While facet joints have been suggested to play an important role in coupling (Oxland et al., 1992a; Panjabi et al., 1989; Shirazi-Adl, 1994b), our results showed that removal of posterior elements had only very limited effects. In fact, only coupling in lumbar spine sections, mainly in AR loading, was substantially affected. Moreover, in both human and porcine spines, reduced coupling percentages during lumbar AR loading appeared to be mainly due to an increase of motion in the primary direction (axial rotation) rather than a reduction in coupled motions (Figs. 2 and 3). This is in line with findings of Heuer et al. (2007). In addition, coupling direction in human lumbar sections changed towards more extension after posterior element removal, which is consistent with previous *in vitro* (Oxland et al., 1992a) and modelling (Shirazi-Adl, 1994b) work.

Porcine spines were, like human spines, only affected by posterior element removal in the lumbar spine. However, relative coupling was substantially smaller in porcine than in elderly human spines. Differences in geometry (Busscher et al., 2010a) and the pattern of flexibility (Wilke et al., 2011) have previously been reported between human and porcine spines. The present study adds that porcine spines are not similar to elderly human spines in coupling behaviour. Regrettably, in the absence of available young human spines, we cannot assess to what extent these differences are either due aging or to species.

Furthermore, we removed the rib cage and this might affect both primary and coupled motions (Mannen et al., 2015). Another limitation of our study is that our spine sections had limited length (4 vertebrae), which may result in underestimation of coupling (Kettler et al., 2000). Furthermore, vertebral motions at the end points were blocked in one (for FE and LB loading) or two (for AR loading) planes. Although the middle segment was not directly restricted, the end-point restrictions could, to some extent, have caused moments in the blocked planes even in the middle intervertebral joints, which could have affected the coupled motions of the

middle segment (van Engelen et al., 2015). This might hamper comparison with previous studies where the endpoints were not blocked. However, in a recent paper (Holewijn et al., 2017) we used the same loads and loading device for much longer thoracic spine sections (T5–T12 for human spines and T3–T10 for porcine spines) and we found very similar magnitudes of coupled motions in the middle 3 (out of 7) motion segments. This suggests that endpoint restrictions and spine section lengths only had minor effects in the present study. Thus, we hypothesize that, with initially globally aligned axes, a fully unconstrained loading device would, compared to our study, only show minor differences (increases) in coupled motions. In addition, we would anticipate in accordance with Heuer et al. (2007), similarly limited effects of posterior element removal.

In addition, to avoid buckling of our multi-segmented spine sections, we did not apply compression during spine testing. Compression, which does occur *in vivo* during spine bending, may increase facet joint contact forces, and thereby affect coupling. Previous testing with follower loads (Wilke et al., 2003) showed about one third reduction of ROM in the loaded direction for the thoracic spine (Sis et al., 2016) and in AR of the lumbar spine (Rohlmann et al., 2001) probably by reducing ‘play in the facet joints’ (Rohlmann et al., 2001). While a follower load could potentially reduce buckling problems with compression, follower loads can never precisely and continuously be applied through the centre of rotation during spine bending, and therefore may cause substantial additional bending moments of unknown magnitude. Moreover, follower loads have been shown to cause substantial shear loads (Cripton et al., 2000).

Furthermore, we loaded human spines at 4 Nm, which is somewhat lower than loads used for the lumbar spines in many other studies. Wilke et al. (1998) recommended 7.5 Nm, but lower loads for high spine levels and fragile spines. As we aimed to use similar loads across the whole spine, and we used elderly spines, our load levels were in accordance with these guidelines. Pilot work for our study indeed showed failures for high spine levels at higher loads.

All of our tests started in neutral position. Panjabi et al. (1989) found that coupling in AR and LB load depended on spine flexion posture, but this may in part have been due to alignment rather than coupling. Furthermore, we investigated human lumbar coupled motions in L2–L3 and not in lower lumbar joints. Oxland et al (1992a) reported L5–S1 coupled motions to be affected by facet joint removal during lateral bending loading, whereas we found effects in L2–L3 during axial rotation loading.

In addition, in accordance with Mannen et al. (2015) we used Euler decomposition, starting with the loaded axis. The order of subsequent coupling axes is, while consistent with clinical intuition, somewhat arbitrary. This order can be disputed, and choosing another order does have effect on coupled motions (Crawford et al., 1996). However, with the rotations in loaded axes below 8 and 18°, respectively, seen for human and porcine spines in the present study, the magnitude of coupling axes order effects is quite limited, especially for the human spines (Crawford et al., 1996). Moreover, such effects are mainly a shift of part of the coupling angle from one coupling axis to the other (Crawford et al., 1996). While this would have some effect on coupling presented in Figs. 2 and 3, it would largely cancel out in our statistical analyses and Fig. 4, which were based on averages over the two coupling axes.

Even when ignoring the issue of axes alignment, comparison to *in vivo* results is difficult. While we used fresh-frozen cadaveric specimen, and even multiple freeze-thaw cycles appear not to have major effects on intervertebral joints (Tan and Uppuganti, 2012), some superhydration cannot be excluded. We used a preloading period to reduce such effects. Moreover, repeated sets of three load cycles and 30 min of creep loading had been applied prior to the present protocol. Note that these creep loads were only half of

the magnitude of the cyclic loads. Furthermore, after recovery, while not being submerged in saline, range of motion in the loaded direction was only marginally different from initial conditions (Busscher et al., 2011). Finally, ligaments had been resected, but this was shown previously not to affect ROM in the loaded directions (Busscher et al., 2011).

In conclusion, the magnitude of coupled motions varied across spine levels and were largest in AR loading of the lumbar spine. Coupled motions were about twice as large in human compared to porcine spines and were hardly affected by posterior element removal. An exception was AR loading of human lumbar and porcine lumbar and low thoracic sections. Here posterior element removal reduced coupling, mainly by an increase of the primary motion, and in human spines this was accompanied by a change towards more extension coupling in both left and right AR loading.

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Conflict of interest

None declared.

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